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This report presents progress achieved through the second 12 months of Phase II of a two phase technology development program.

During Phase I, Lockheed Martin demonstrated ultrasound imaging at 5 MHz using an electronically scanned, two-dimensional hybrid transducer array and acoustical lens. The hybrid array was constructed by mating a readout integrated circuit (ROIC) originally designed for infrared imaging to a 44x64 element piezoelectric array.

Thus far in Phase II, developments include: a new ROIC specifically designed for ultrasound imaging; a 128x128 Transducer Hybrid Assembly (THA), transmitter electronics and transducers, a wide-angle diffraction limited acoustical lens and ancillary electronics for readout, signal processing and display. The design of these subsystems was described in the 1996 Annual Report. They are now functional and system integration is underway. Initial results with water tank imaging of simple targets are presented in this report.

The laboratory demonstration system being developed will produce real-time (30Hz), two-dimensional images with 1 mm resolution in three dimensions and will be capable of collection of three-dimensional data for real-time volume imaging.

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Timothy E White

PI - Signature Date

INTRODUCTION

An acoustical camera (Fig. 1) is under development in this program for ultrasound imaging of abdominal trauma. The system is intended for rapid evaluation of battlefield injuries by paramedical personnel. With a real-time, volumetric (3 dimensional) capability, a portable, "point-and-shoot" approach is expected to be feasible. This will bypass the need for the time-consuming, detailed manual scanning/assessment which is characteristic of current ultrasound systems.

The camera is the direct analog of a conventional video camera. It requires a large area, high pixel density, fully populated, two-dimensional (2D) transducer array to act as the sensing element in the focal plane of an acoustical lens. Conventional approaches to two-dimensional array fabrication have been constrained by the technique employed to interconnect the transducer and the initial stage of electronics. These approaches rely on micro-coaxial cable soldered to each array element. Although micro-coax technology has improved dramatically in the past decade, interconnecting thousands of array elements with separate wires remains a formidable challenge. In addition to this practical fabrication issue, the capacitance of a long coaxial cable (typically 40 pF/m) is much larger than that of a typical 2D array element (< 1pF). This creates a voltage divider that severely reduces the signal-to-noise ratio of the channel.

In Phase I of this program, Lockheed Martin IR Imaging Systems in 1995 demonstrated such a two-dimensional transducer array. It used a "flip-chip" or direct interconnection approach to join a 2D composite transducer array to a silicon readout integrated circuit. The direct interconnect technique allows array elements to be individually connected en-masse. Furthermore, the length of the direct interconnection is less than 0.1 mm, reducing interconnection capacitance to the level where it is no longer a dominant factor in the channel signal-to-noise ratio.

This interim report for Phase II of the program, describes progress in the design and construction of a 5 MHz imaging system based on a 128 x 128 THA with 0.2 mm array center to center spacing.

The goals of our medical imaging system development are three-dimensional visualization of the interior of a human subject with:

- Excellent image quality
- Real-time frame rates
- Image manipulation to obtain any derived view of cross-section
- Portability
- Low cost

The ultimate goal of the acoustical camera development is to provide a manportable ultrasound imaging capability for rapid evaluation of battlefield injuries, particularly abdominal trauma. By moving a capability to monitor internal bleeding to forward echelons, triage decisions may be

optimized in the "golden hour" so vital for survival. Compared to conventional B-scan ultrasound systems which only image a single plane at a time and require mechanical (typically motorized) scanning for 3D coverage, the ability to simply "point and shoot" with the acoustical camera will make imaging more reliable under the often chaotic conditions of the battlefield. Furthermore, transmitting these images to a radiologist at a field hospital will allow a diagnosis to be made in a more benign environment. In many cases, additional imaging may not be needed when the individual reaches the field hospital and they may proceed rapidly to the next phase of therapy.

Our acoustical camera is the application of dual use technology derived from other DARPA sponsored programs. Arrays are derived from IRFPA technology, using acoustically sensitive materials in place of the IR detectors. Under the flexible manufacturing program with DARPA/MTO, Lockheed Martin IR Imaging Systems has also demonstrated fabrication of ultrasound Transducer Hybrid Assemblies (THA's). By using the same IR hybrid fabrication line that supplies DOD and commercial IR imaging products, ultrasound arrays have been made available to this program without significant startup expenses, providing maximum synergy of DARPA programs.

We have assembled a team that includes key leaders in diagnostic ultrasound medicine, equipment development and imaging systems:

University of Rochester Center for Biomedical Ultrasound:

Kevin Parker, Ph.D., Director. The center is the oldest and largest interdisciplinary research organization in the world devoted to all aspects of medical imaging.

Deborah Rubens, MD Dr. Rubens is a world class radiologist and expert in ultrasound imaging. She will lead the U of R team evaluating the acoustical camera.

Lockheed Martin IR Imaging Systems:

A leader in military and commercial applications of electro-optic imaging technologies, LMIRIS is designing, fabricating and testing the acoustical camera.

We have on-going discussions with a number of companies in the medical ultrasound imaging field. It is Lockheed Martin's intention to enter into an OEM agreement with a major medical imaging system manufacturer or distributor to bring the camera to market.

The following sections describe our progress towards these goals. The technical and scientific merit of our work has been demonstrated through one invited lecture, several presentations at scientific/engineering society annual meetings and three publications.

TECHNICAL ACCOMPLISHMENTS:

ARRAY DESIGN AND FABRICATION: A 1-3 composite piezoelectric material of 45% piezoceramic volume fraction was chosen for the ultrasound array. This yields elements of improved sensitivity, with a typical capacitance of several hundred femtofarads (fF), and remains within the practical limits of present fabrication techniques. The center frequency of the array elements is 5.0 MHz with a center-to-center spacing of 0.2 mm chosen to achieve spatial sampling of approximately one wavelength. The elements were delineated by sawing through one electrode layer. A common ground layer was used on the matching layer side. 16x16 and 32x32 test arrays and 128x128 arrays were fabricated and tested.

Several 32x32 arrays were mounted on a lead-out board to provide electrical access to individual array elements. Measurements of array performance, including center frequency, bandwidth and sensitivity were performed on these arrays confirmed the acoustical design parameters.

INITIAL THA MEASURED PERFORMANCE: A 16x16 element 3.0 MHz test array was hybridized to a single ROIC for initial testing. The average measured sensitivity was -194 (±2) dB re 1V/μPA with a broadband electronic noise level of 84 μV. Using relatively uniform direct insonification, Fig. 2 demonstrates the range resolution of the THA by placing a short acoustical pulse in one range plane.

Several 128x128 THA's have been fabricated, however, at present yields in the hybridization process for these larger arrays are not acceptable. Ongoing process development in the next year is expected to improve these yields and to produce an acceptable THA for initial in-vitro imaging.

ACOUSTICAL LENS DEVELOPMENT: A wide angle acoustical lens (Fig. 3) has been designed, fabricated and tested. Figure 4 is an image taken with this lens by scanning the lens and target together as a unit over a single element of an array. This is equivalent to electronically scanning with the array and a stationary lens and target. Although this image was made at 2.8 MHz, with a substantially longer wavelength, the resolution and detail in the image demonstrates the excellent diffraction limited performance of the lens. At 5 MHz, this lens will provide the 1 mm resolution to meet one of the key goals of the program.

CONCLUSIONS

Lockheed Martin IR Imaging Systems has demonstrated the basic functionality of the THA design and shown that the initial versions meet the design goals. All other components of the in-vitro imaging system are complete and in system integration awaiting an improved THA. Fabrication and testing of additional THA's is underway. In the next year, we expect to achieve real-time in-vitro imaging.

REFERENCES:

1996 Annual Report - DAMD17-94-J-4511

APPENDICES:

Appendix A: Ken Erikson et al, "128x128(16k) Ultrasonic Transducer Hybrid Array", to be published in *Acoustical Imaging*, Vol. 23, ed. S. Lees, Plenum Publishing Corporation, New York.

Appendix B. Ken Erikson et al, "A 128 x 128 (16k) Ultrasonic Transducer Hybrid Array", to be published in *Proc.1997 IEEE Ultras. Symp.*, 1998.

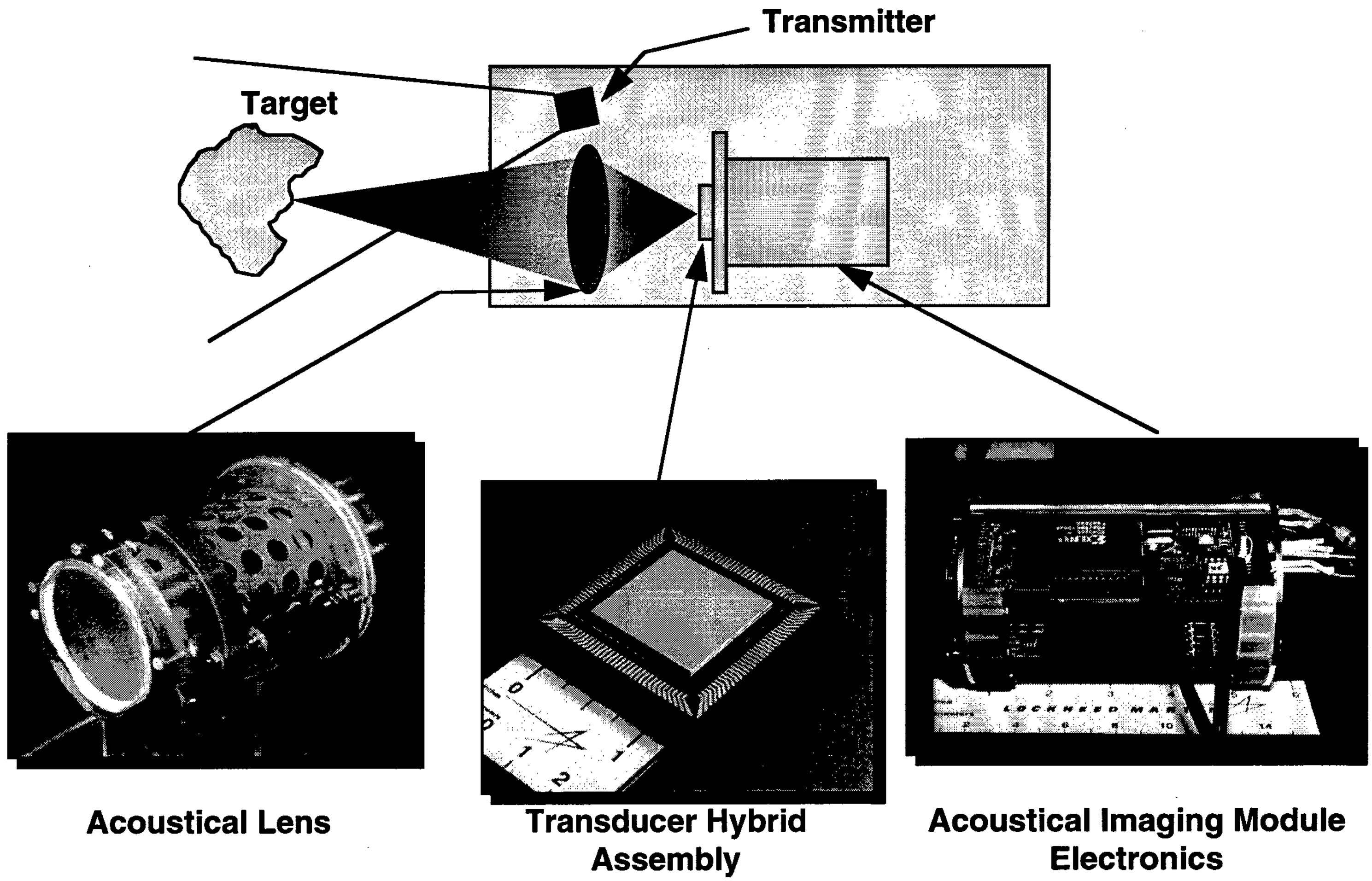


Figure 1. Acoustical Camera Components

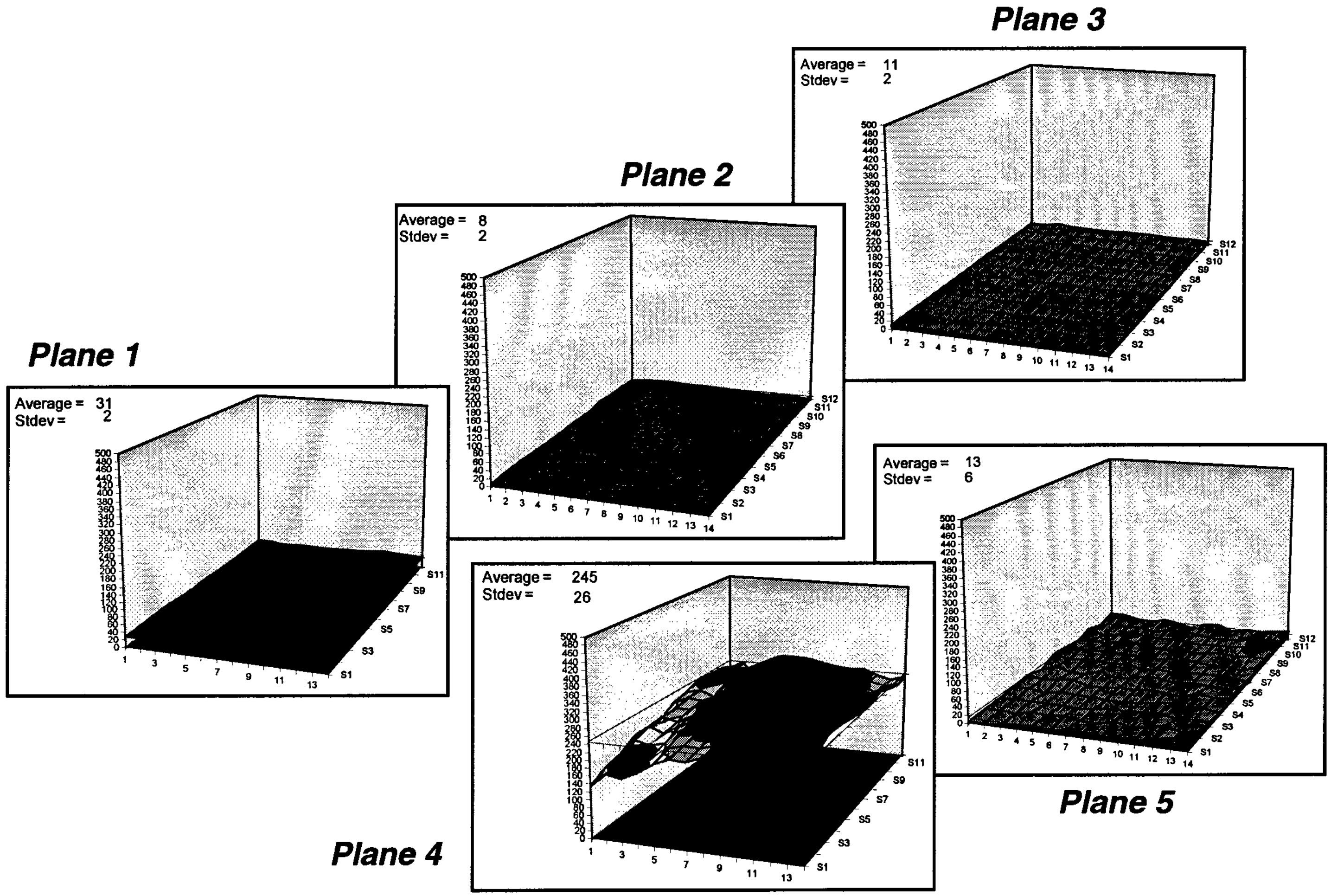
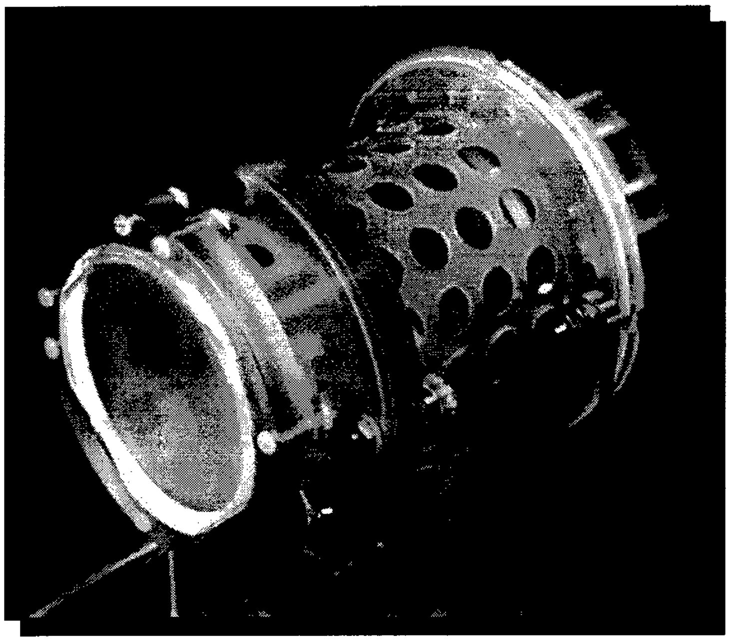


Figure 2. THA Range Resolution. A Short Acoustical Pulse (4.0 ms) Appears Only in Range Plane 4. The Range Gates are 3.25 ms Long and are Spaced by 0.75 ms, Corresponding to Pulse-Echo Ranges of 2.4 mm and 0.5 mm Respectively



Nominal Operating Frequency:	***5 MHz***
Focal Length:	***72 mm***
Entrance Pupil:	***77 mm***
f/#:	***1.2*** ***(@ used conjugates)***
Object Depth:	***0 cm - 15 cm***
Field of View:	***40 °***
Object Size at Depth:	***8 cm x 8 cm***
Magnification:	***0.32***
Blur Size:	***1.08 mm*** ***(85% Encircled energy)***

Figure 3. Phase II Acoustical Lens

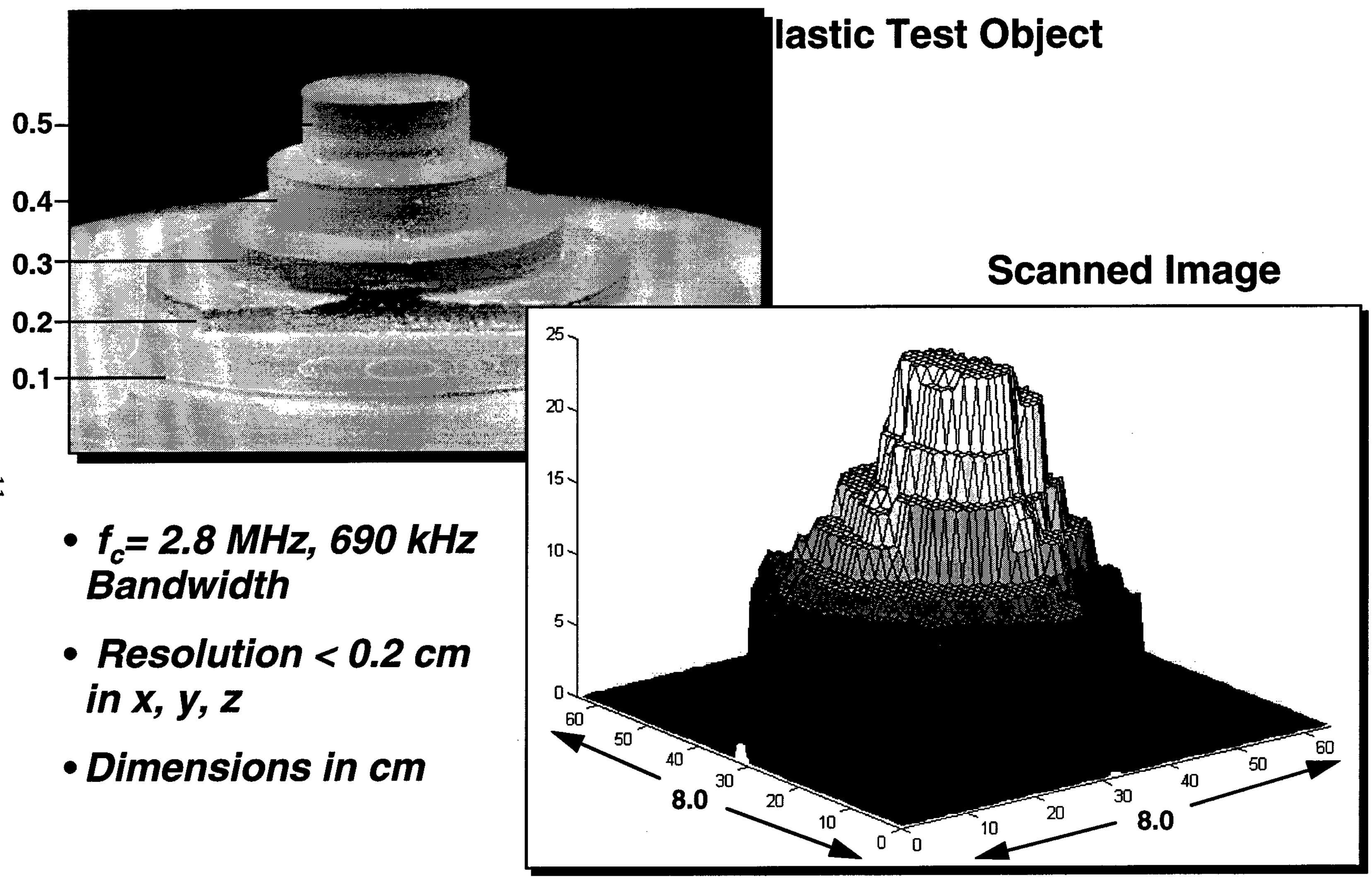


Figure 4. Acoustical Image Obtained With Lens of Figure 3

APPENDIX A
To be published in:
Acoustical Imaging, Vol. 23, Ed S. Lees,
Plenum Publishing Co., N.Y., 1998.

A 128 X 128 (16k) ULTRASONIC TRANSDUCER HYBRID ARRAY

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INTRODUCTION

Ultrasonic imaging in the low MHz frequency range with large two dimensional arrays presents many challenges in design and fabrication. In this paper, a 128 x 128 (16,384 total) element receiver array, consisting of a 1-3 composite piezoelectric transducer array bonded directly to large custom integrated circuits is described. This Transducer Hybrid Array (THA) is intended for use in a real-time 3D imaging system or acoustical camera (Fig. 1) for medical and underwater applications.

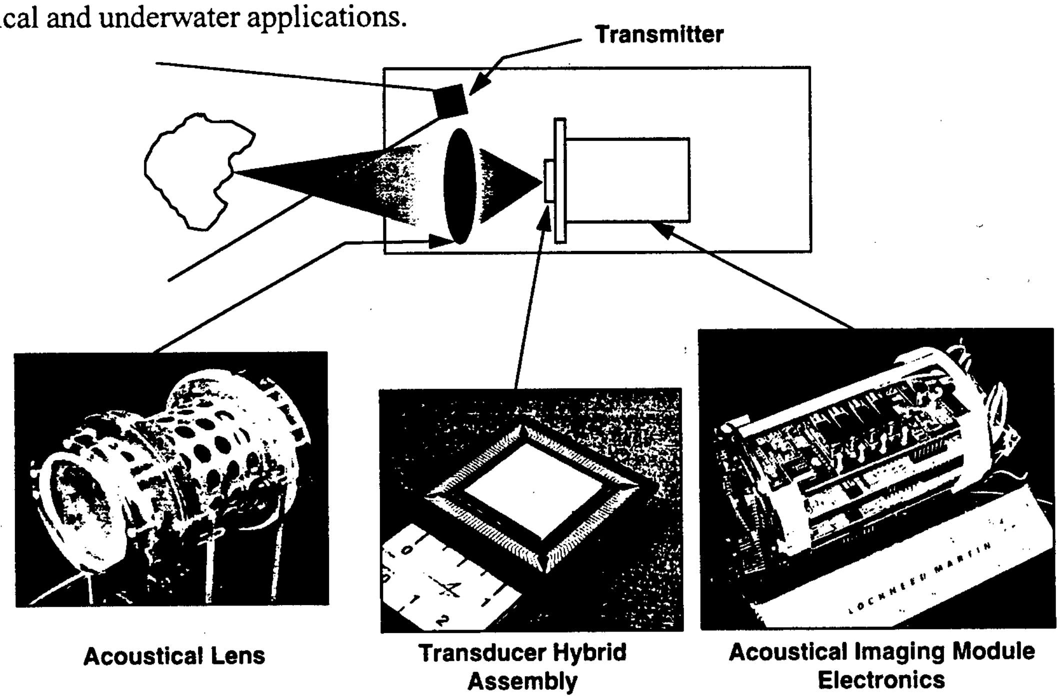

Figure 1. Real-time 3D Acoustical Camera using the Transducer Hybrid Array.

Traditional ultrasonic imaging systems, whether for medical or underwater use, are also operated in a monostatic mode, i.e., the same array element is used for both transmission and reception. This poses practical difficulties in fabricating an integrated circuit with both the high density required for massively parallel on-chip signal processing and the high voltage capabilities required for adequate transmitted power. The bistatic mode, where the transmitter is separated from the receiver was selected for the array discussed here. Although the THA was first demonstrated[1] over 20 years ago, advances in microelectronics technology have now made such a device practical.

Conventional ultrasound systems use micro-coaxial cable to connect the array to the front end electronics. Although micro-coax technology has improved dramatically in the past decade, interconnecting 16,384 array elements with separate wires remains a formidable challenge. In addition to this practical fabrication issue, the capacitance of a long coaxial cable (typically 40 pF/m) is much larger than that of a typical 2D array element (< 1pF). This creates a voltage divider that severely reduces the signal-to-noise ratio of the channel. The direct connection method used in the THA (Fig. 2) reduces the interconnect length to less than 0.1 mm, reducing interconnection capacitance to the level where it is no longer a dominant factor in the channel signal-to-noise ratio.

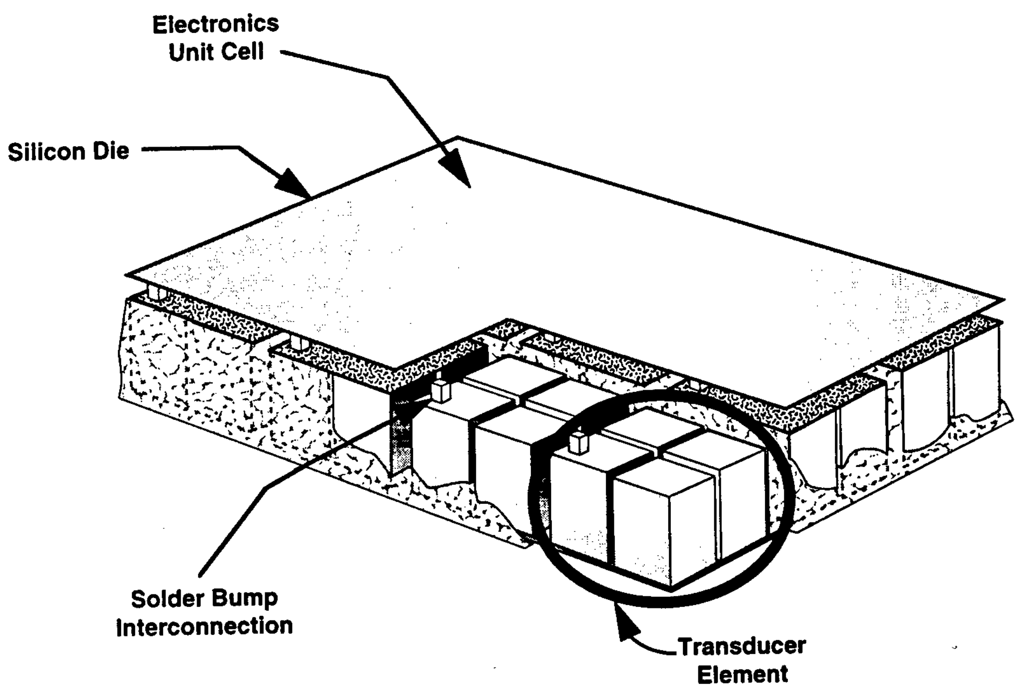

Figure 2. Cross-section of Transducer Hybrid Array showing the interconnection method.

Finally, real-time 3D imaging imposes additional severe requirements on an acoustic array and signal processing system. Due to the finite velocity of sound, ranges greater than a few mm in tissue or water require either multiple parallel beam forming in a B-scan system or C-scan parallel data acquisition.

Conventional B-scan systems acquire data one line at a time, typically waiting until well after information from the furthest range has been received before transmitting the next pulse. With single plane imaging, this is not a serious limitation. Indeed, real-time imaging is one of the key strengths of present day medical ultrasound.

3D Ultrasound (3DUS) has been gaining increasing acceptance in medical and underwater imaging applications due to the improved image understanding it provides and to the additional applications it enables. 3DUS systems have been developed using conventional B-scan probes mechanically scanned in the third dimension. The B-scan images are subsequently assembled into 3D volume images off-line. To scan out a volume with comparable resolution in the third dimension, however, results in very low frame rates. For example, to scan an 80 mm cubic volume centered at 120 mm with 1 mm^3 resolution requires 6,400 separate pulses, which in turn requires well over one second for data acquisition under the most favorable assumptions.

Various strategies[2] have been devised to improve the frame rate of B-scan ultrasound systems. Improvements of a factor of 3 or 4 may be achieved; however, this is still far from real-time. Another successful approach[3] uses a wide insonification beam together with multiple parallel receive beamformers to present a somewhat lower resolution image at real-time rates sufficient for cardiac imaging.

A C-scan format (Fig. 3) was chosen for the present system to enable real-time 3D imaging.

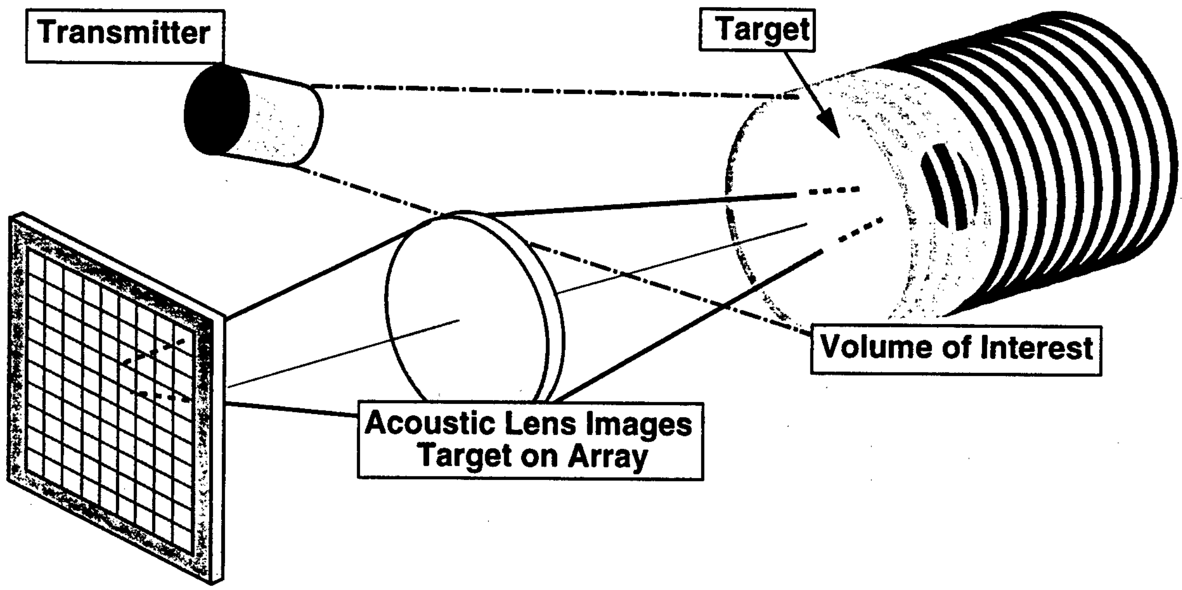

Figure 3. C-Scan "Plane at a Time" imaging enables real-time 3D operation.

With this C-scan, bistatic system a separate pulsed transmitter insonifies the entire volume of the target. An acoustical lens images successive planes in the target onto the acoustical array. As shown in Fig. 3, five range planes can be recorded from a single pulse of sound. By simply delaying the range gate timing on the next pulse, the next five planes in the target are captured, subject to depth of field limitations of the acoustical lens.

The acoustical frame rates enabled by this system are very high. For example, to record data from an 80 mm cubic volume centered at 120 mm range with a 1 mm^3 resolution results in an acoustic frame rate of approximately 300 Hz. This is more than enough to support a 30 Hz real-time display, so the extra frame rate may be applied to other signal processing techniques such as frame averaging for noise reduction.

TRANSDUCER HYBRID ARRAY (THA) REQUIREMENTS

The THA is the heart of the real-time 3D system in development. Table 1 lists the performance requirements for the device and are derived from the system performance goals.

Table 1. Performance Requirements for the Transducer Hybrid Array (THA)

Number of elements:	Medical - 128 x 128 (rectangular matrix)
	Underwater - 64 x 64
Element size:	< $\lambda \times \lambda$ square (λ = wavelength in water)
Aperture:	25 mm x 25 mm minimum
Ultrasound frame rate:	$\geq$ 200 Hz maximum
(Supports frame averaging for a 15-30 Hz display update rate)	
Acoustic performance:	
Center frequency:	Medical - 5 MHz
	Underwater - 2.5 - 3.5 MHz
Fractional bandwidth:	> 40%
Sensitivity:	> -200 dB re 1V/µPa
Crosstalk:	< - 30 dB
On-chip signal processing:	
a. Amplifier overvoltage protection	
b. Gain: x2, x20, x200	
c. Maximum Sampling frequency: $\geq$ 20 Megasample/sec	
d. Type of sampling:	
(1). Sample and hold (20 data points)	
(2). Quadrature detection:	
Supports sequential acquisition of 5 planes (range gates) per acoustic pulse.	
Electronic Performance:	
Frequency response:	Bandpass, 150 kHz to > 8 MHz.
Instantaneous dynamic range:	> 60 dB
Noise level	< 50 nV/$Hz^{1/2}$ (referred to input)
Crosstalk:	< - 50 dB
Fixed pattern noise:	< 0.5 dB standard deviation re average level
(Due to pixel to pixel electronic gain and offset variations)	
Total harmonic distortion:	-60 dB @ 2 V output swing
Power:	< 1 watt total
Size (including package and leads):	< 50 mm x 50 mm x 10 mm

ARRAY DESIGN

A single square of 1-3 composite piezoelectric material is used for the ultrasound array[4]. For the 5 MHz medical imaging application, a 0.2 x 0.2 mm center-to-center spacing is used together with a single matching layer. The underwater imaging application uses 2.5 to 3.5 MHz elements on 0.4 mm centers. The elements were delineated by sawing through one electrode layer. A common ground layer was used on the matching layer side. Figure 4 is a photograph of a typical 64 x 64 element array and Fig. 5, a single element. The elements are typically 45% piezoceramic volume fraction and have a capacitance of several hundred femtofarads (fF).

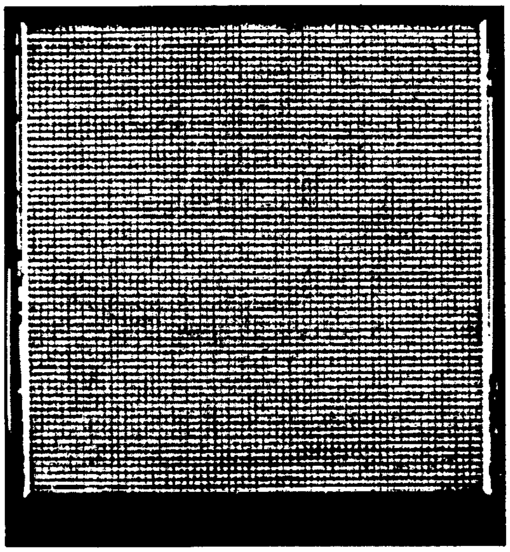

Figure 4. 2D Transducer Array[2]: 25.6x25.6 mm.

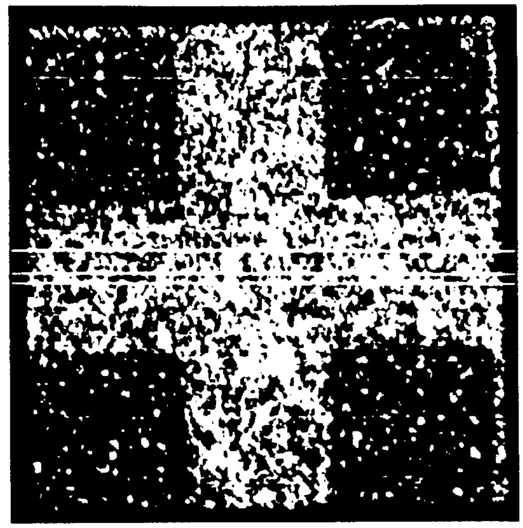

Figure 5. Typical 2.5 MHz array element: 45% volume fraction on 0.4 mm centers.

INTERCONNECTION

The interconnection method, a variation of flip-chip bonding is shown in Fig. 2. Solder bumps are deposited on both the array and the silicon integrated circuit. During hybridization the bumps are optically aligned and brought into contact to make an electrical connection. The contact area of the bumps on the array is approximately 20 x 20 microns, which provides adequate mechanical integrity, good electrical contact and has a minimal (but not negligible) effect on the acoustics of this essentially air-backed transducer design.

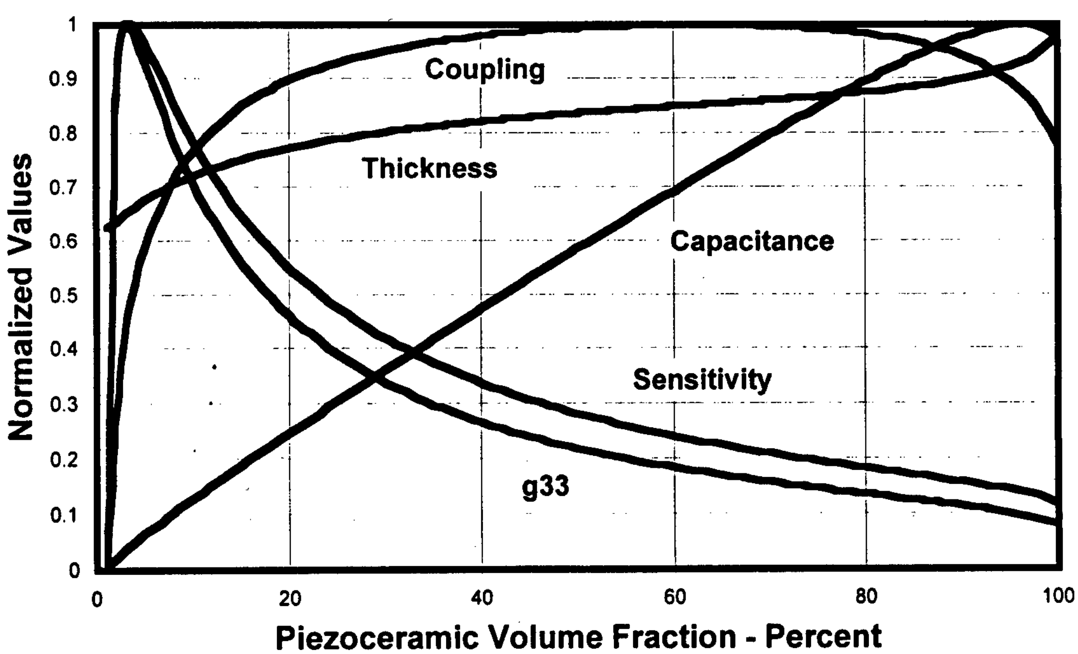

Figure 6. Normalized 1-3 Piezoelectric Composite Parameters *at Constant Resonance Frequency.*

With this interconnection method, the input capacitance of the electronics is less than 100 fF. This provides a unique opportunity to optimize the composite piezoelectric material that is generally not available to conventional ultrasound systems.

Figure 6 plots piezoelectric coupling coefficient, capacitance and sensitivity at *constant resonance* frequency as a function of piezoceramic volume fraction[5]. By reducing the volume fraction of ceramic, sensitivity can be increased at the expense of lower capacitance while maintaining high coupling.

Figure 7 shows the frequency response of a typical element of a 16x16 test array. Sensitivity is excellent and the bandwidth acceptable.

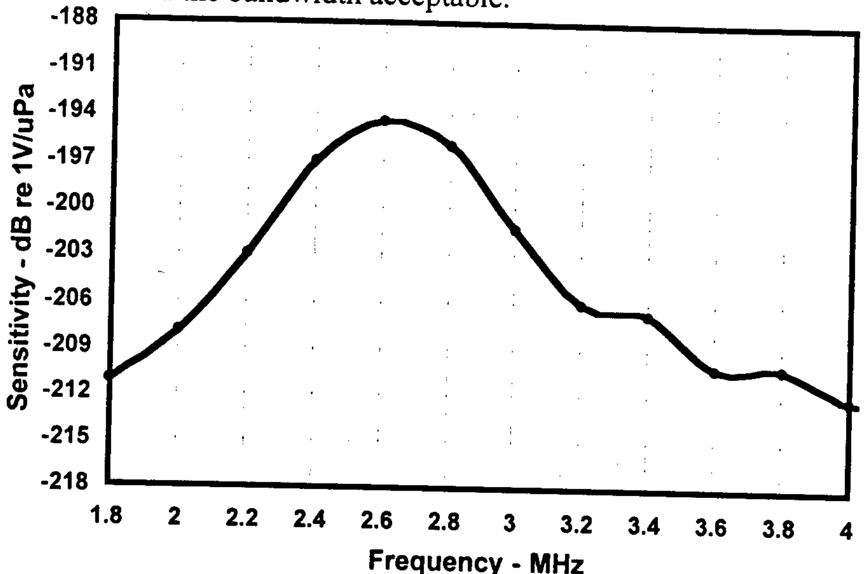

Figure 7. Frequency response of typical array element without an optimized matching layer. The sensitivity was measured with a calibrated PVDF membrane hydrophone. The measured -6 dB bandwidth is 700 kHz.

READOUT INTEGRATED CIRCUIT (ROIC) DESIGN

From a silicon foundry perspective, matching the silicon unit cell to the array dimensions would result in very large die. A four-chip hybrid was selected to keep the IC size reasonable. To avoid a row of dead pixels in the center of the array, the silicon is designed to be butted along two adjacent edges, with all bond pads on the other two edges. The resulting die is 15.5 mm x 15.5 mm and contains 64 x 64 (4,096 total) unit cells together with multiplexing circuits for readout. There are about 1 million active devices per die.

Figure 8 is a schematic diagram of the ROIC. In each unit cell, amplification, filtering, analog sampling and storage of the output of all the transducer elements in the array occurs simultaneously (in parallel) during data acquisition periods (range gates) which are typically a few microseconds long. After data acquisition, the stored data are read out through four 10 MHz pixel rate outputs.

Two modes of operation are possible: sample-and-hold and quadrature sampling. In the sample-and-hold mode, the received waveform is simply sampled at a 20 Megasample/sec maximum rate when the range gate is activated. In the quadrature sampling mode, the waveform is sampled at a rate 4 times the acoustic center frequency (20 Megasamples/s maximum) and stored. Through this well known technique[6], the in-phase and quadrature components of all 16,384 waveforms are simultaneously recorded at 5 separate range gate times. Following data acquisition, the stored data are read out to the system through four outputs per ROIC for a total of 16 readout channels.

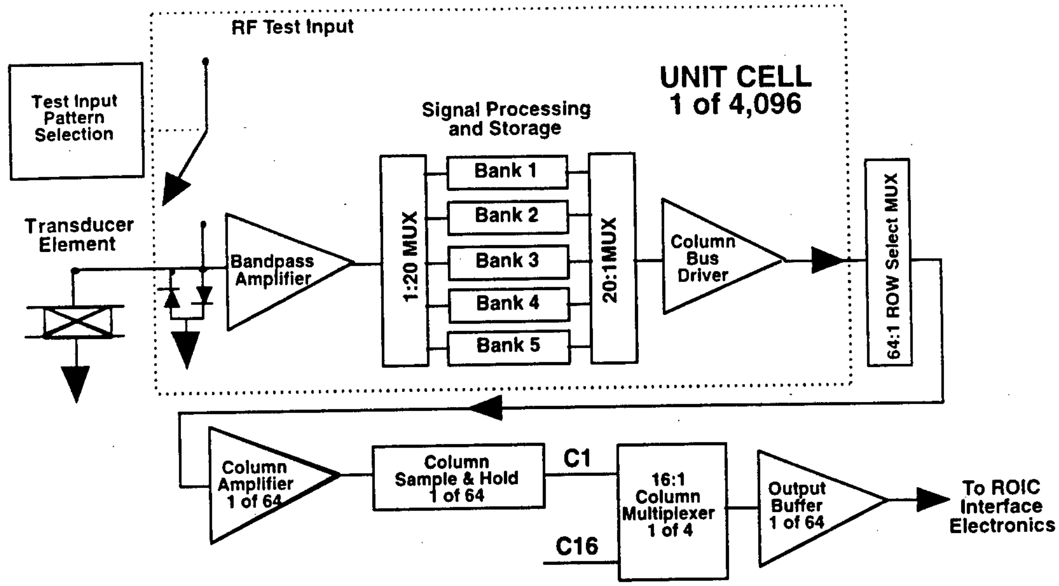

Figure 8. Functional Block Diagram of the ROIC.

INITIAL MEASURED PERFORMANCE

A 16x16 element 3.1 MHz test array was hybridized to a single ROIC for initial testing (Fig. 9). The average measured sensitivity was -188 (±6) dB re 1V/μPA with a broadband electronic noise level of 84 μV. Using relatively uniform direct insonification, typical responses were measured as shown in Figs. 10-13. In these figures, the voltage response of each element is plotted as a surface over a 2D grid representing the element location. The lower response in the corners of the array was found to be caused by a matching layer defect. In addition, for this initial test a less than perfect ROIC was used, resulting in some non-functional rows and columns which are omitted from the plots.

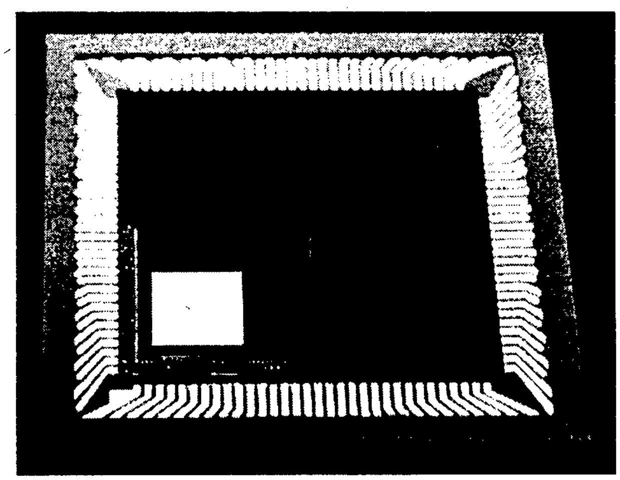

Figure 9. A 16x16 test array hybridized to a single ROIC for initial testing.

With a 60 μs long gated sine pulse, activity may be seen in each of the 5 planes (Fig. 10). Figure 11 demonstrates the range resolution of the THA by placing a short acoustical pulse in one range plane. Without any acoustical input (Fig. 12), the "fixed pattern" of the THA

may be observed. Since these gain and offset variations are constant for each element of the array, they may be compensated for later in the system electronics. Finally, the system noise level is demonstrated in Fig. 13, where the results of subtracting data from the same plane in two successive frames is presented. This is equivalent to having corrected the fixed pattern variations, leaving only the random electronic noise.

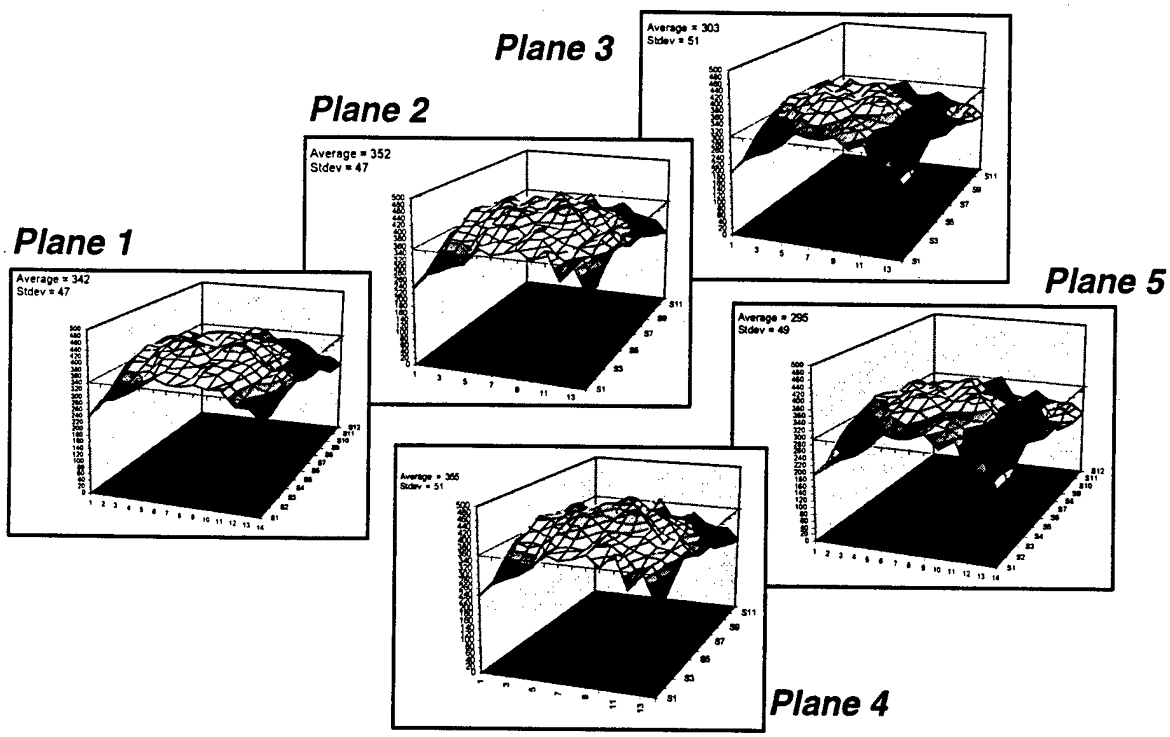

Figure 10. Response of the test THA to a 60 µsec gated sine pulse of low amplitude ultrasound. The range gates were set to capture the full acoustical pulse.

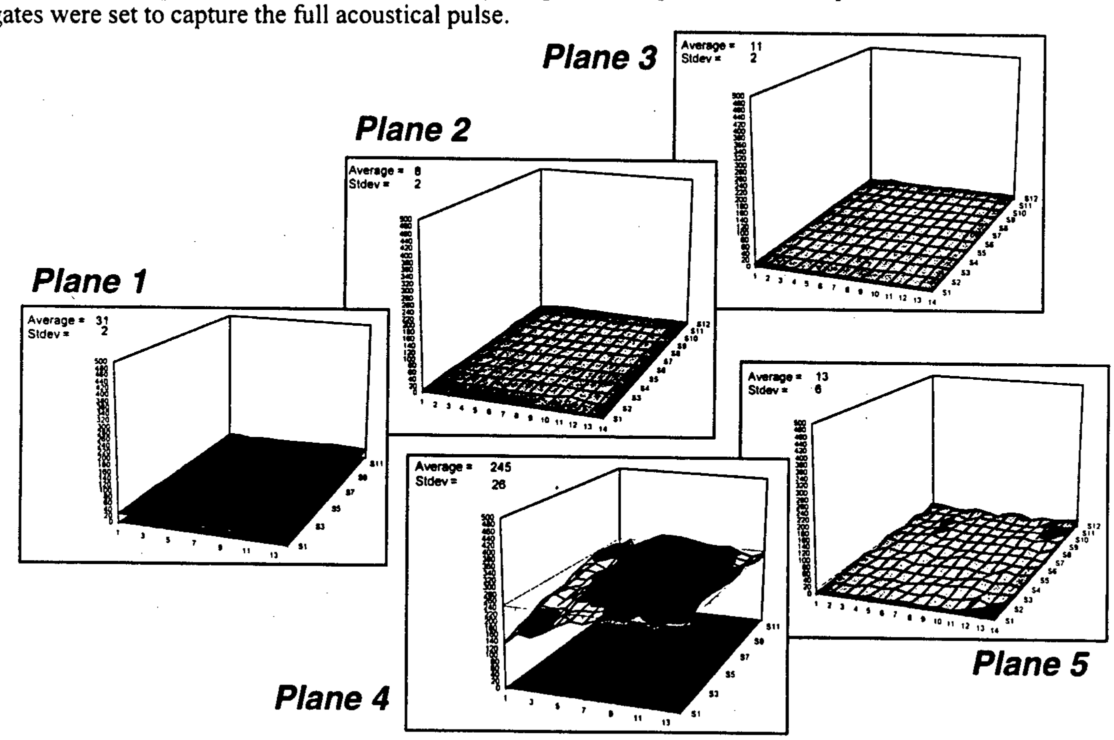

Figure 11. THA Range resolution. A short acoustical pulse (4.0 µs) appears only in range plane 4. The range gates are 3.25 µs long and are spaced by 0.75 µs, corresponding to pulse-echo ranges of 2.4 mm and 0.5 mm respectively.

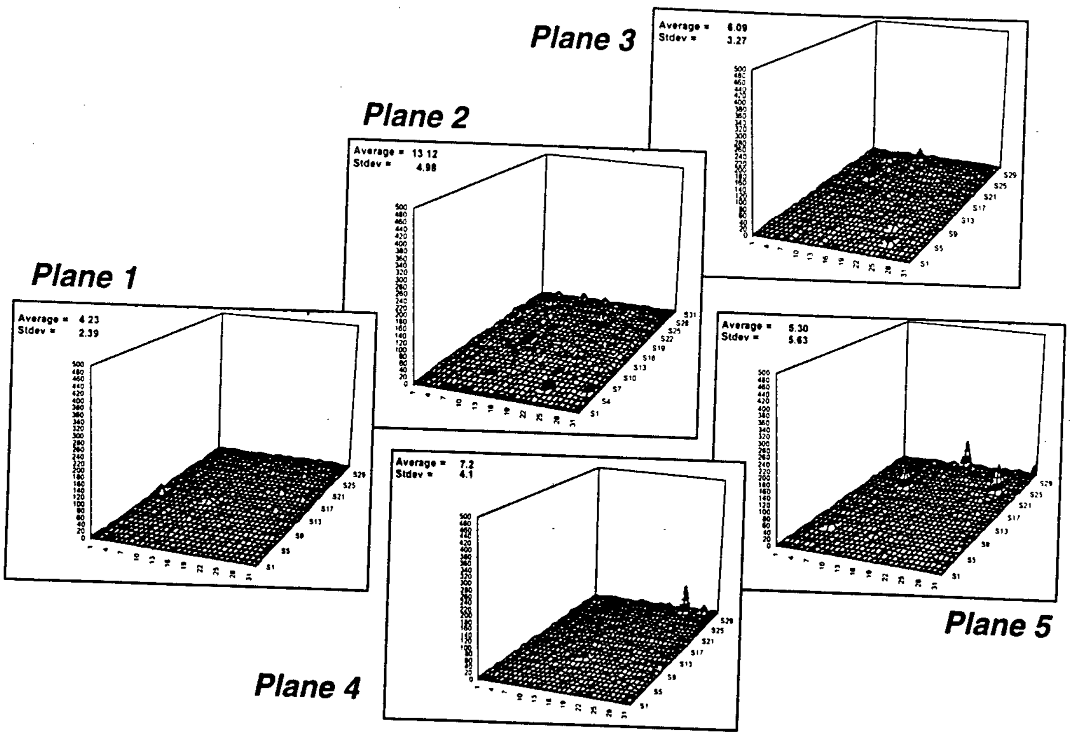

Figure 12. With no acoustical input, the "fixed pattern" noise of the 16x16 THA is revealed.

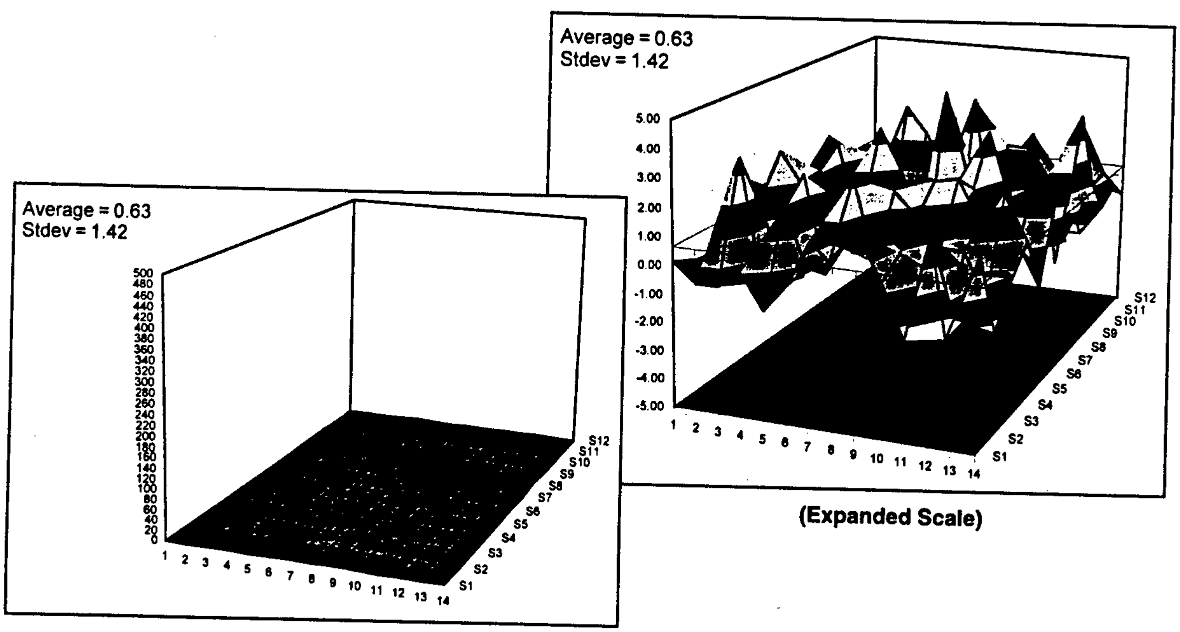

Figure 13. With no acoustical input, THA frame-to-frame repeatability is demonstrated by subtracting the same plane in one frame from the next frame. The difference is equal to the system electronic noise level. The right plot (b) is an expanded version of the left plot (a).

CONCLUSION

Fabrication and testing of 64x64 and 128x128 THA's is underway. Figure 14 is a schematic drawing of the THA, and Fig. 15 is a photograph of one such large array in the PGA package.

We have demonstrated the basic functionality of our THA design and shown that the initial versions meet the system performance goals.

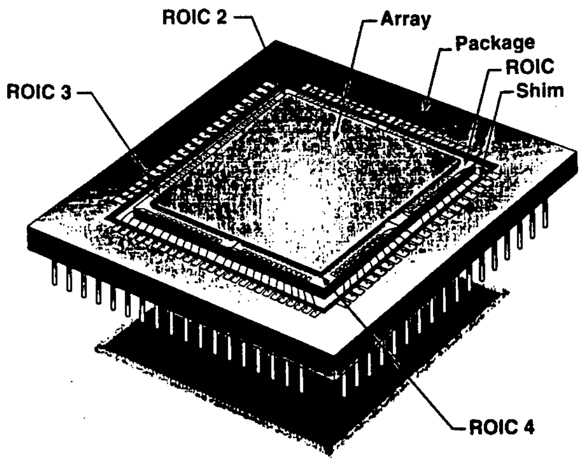

Figure 14. Schematic drawing of the 64x64 (or 128x128) THA.

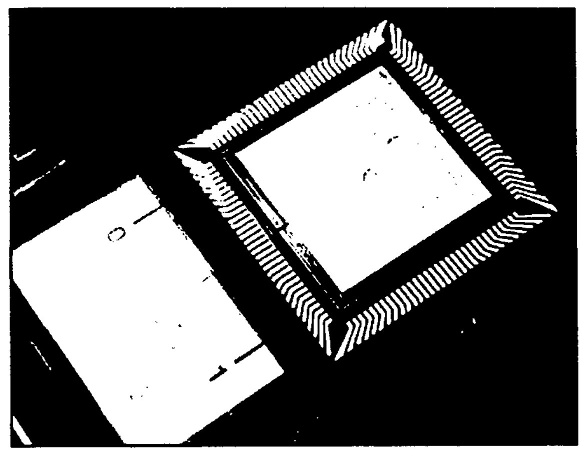

Figure 15. A 64x64 element 3.1 MHz THA.

ACKNOWLEDGMENTS

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REFERENCES

1. K. Erikson and R. Zuleeg, "Integrated Acoustic Array," *Acoustical Holography*, Vol. 7, ed. L. Kessler, Plenum Pub. Corp., pp. 423-444, 1977.
2. K. Erikson, S. Ishrak, W. Reckwerdt and R. Spratt, U.S. Patent No. 5301674, April 12, 1994.
3. O. Von Ramm and S. Smith, U.S. Patent No. 4694434, September 15, 1987.
4. American Medical Design, Paso Robles, CA, 93446
5. W. A. Smith and B. A. Auld: "Modeling 1-3 Composite Piezoelectrics: Thickness-Mode Oscillations," *IEEE Trans. Ultrasonics, Ferroelectrics and Frequency Control*, **UFFC-38:** 40-47 (1991).
6. D. Horvat, J. Bird and M. Goulding, "True Time-Delay Bandpass Beamforming," *IEEE J. Oceanic Eng.*, 17: 185-192 (1992).

APPENDIX B
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A 128 X 128 (16k) ULTRASONIC TRANSDUCER HYBRID ARRAY

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Abstract - Ultrasonic imaging in the low MHz frequency range with large, dense arrays presents many design and fabrication challenges. Conventional ultrasound systems use micro-coaxial cable to connect the array to the front end electronics. While coax technology has improved dramatically in the past decade, interconnecting 16,384 array elements with separate wires remains a formidable challenge. In addition to this practical issue, the capacitance of a long coaxial cable (~40 pF/m) is much larger than that of a typical 2D array element (< 1pF), creating a voltage divider that severely reduces the signal to noise ratio of the channel.

A 2D composite piezoelectric receiver array bonded directly to four large custom integrated circuits is described. This 128 x 128 (16,384 total) element Transducer Hybrid Array (THA) of 200 μm unit cell spacing is intended for a 3D real-time imaging system for medical and underwater applications. By reducing the interconnect length to less than 20 μm, cable capacitance is no longer a problem. Massively parallel, on-chip signal processing enables true real-time three-dimensional imaging. Favorable tradeoffs using composite piezoelectric materials, enabled by this high-density flip-chip interconnection technology are discussed.

I. INTRODUCTION

Two dimensional arrays remain one of the last frontiers of ultrasonic imaging with real-time three-dimensional imaging as the ultimate goal. 3D Ultrasound (3DUS) is gaining acceptance in medical and underwater imaging applications due to the improved image understanding it provides and to the additional applications it enables.

3DUS systems have been developed using conventional B-scan probes mechanically scanned in the third dimension. The B-scan images are subsequently assembled into 3D volume images off-line. Scanning a volume with such a system with comparable resolution in three dimensions, however, results in very low frame rates. For example, to scan an 80 mm cube centered at 120 mm with 1mm^3 resolution requires 6400 separate pulses, which in turn requires well over one second for data acquisition.

In this paper, we present the design and initial experimental results with a large, fully populated Transducer Hybrid Array (THA) which is the enabling technology for a true real-time 3DUS system or acoustical camera (Fig. 1). Although the THA was first demonstrated[1] over 20 years ago, advances in microelectronics technology have now made such a device practical.

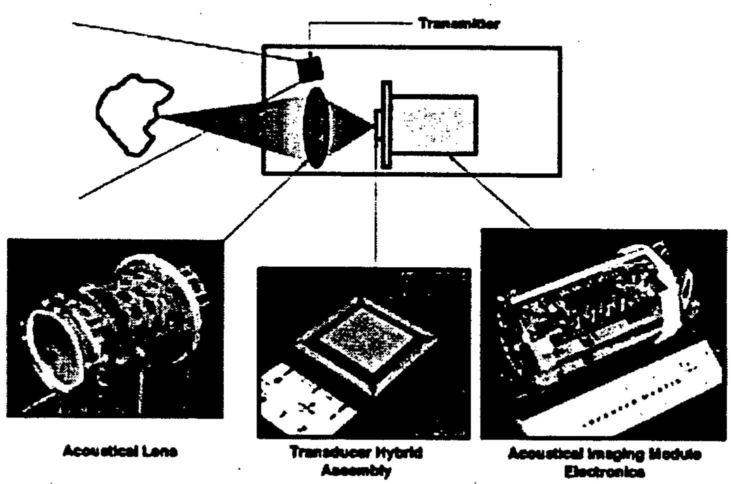

Fig. 1. Real-time 3D Acoustical Camera using the Transducer Hybrid Array

This acoustical camera is a direct analog of an optical video camera. A transmitter insonifies a target and reflected energy is focused as an image by a multi-element acoustical lens. A THA in the image plane of the lens serves as an all electronic readout, with no moving parts. The acoustical imaging module electronics controls the THA, generates the transmitter signals, digitizes and preprocesses the returns

The THA operates as a parallel read-in device, i.e. all 16,384 input channels are simultaneously active, amplifying, processing and storing 5 complete planes of amplitude and phase information *per acoustic pulse* in a C-scan format. This parallel read-in capability is the key to real-time 3D operation. To capture a volume of data with this "plane-at-a-time" capability (Fig. 2), the range gates are delayed from pulse to pulse, sweeping out a volume only limited by the depth of field of the lens.

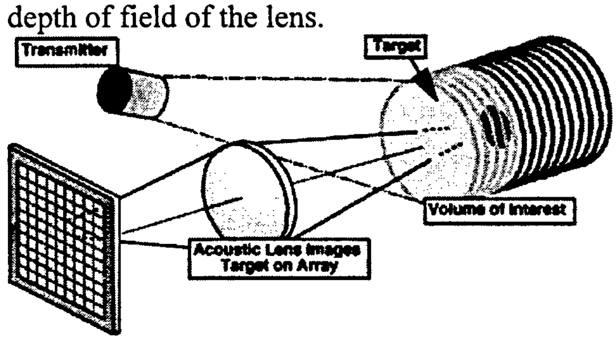

Fig. 2. C-Scan "Plane at a Time" imaging enables real-time 3D operation

II. THE TRANSDUCER HYBRID ARRAY

The Transducer Hybrid Array (THA) is composed of a custom silicon CMOS integrated circuit, flip-chip bonded directly to a composite piezoelectric array (Fig. 3).

Acoustical Design

A single square of 1-3 composite piezoelectric material is used for the ultrasound array[2]. For the 5 MHz medical imaging application, a 0.2 x 0.2 mm center to center spacing is used together with single matching layer. The underwater

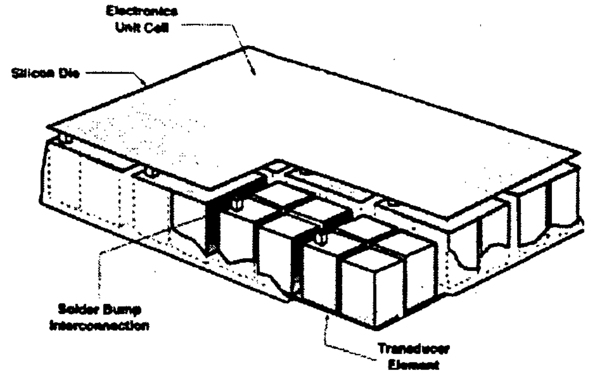

Fig. 3. Cross-section of Transducer Hybrid Array showing the interconnection method.

imaging application uses 2.5 to 3.5 MHz elements on 0.4 mm centers. The elements were delineated by sawing through one electrode layer. A common ground layer was used on the matching layer side. The elements are typically 45% piezoceramic volume fraction and have a capacitance of several hundred femtofarads (fF). Bandwidths of 700 kHz at a 2.6 MHz center frequency and sensitivities of -190 dBV/mPa are routinely achieved.

Interconnection

Complementary solder bumps (Fig. 3.) are deposited on the array and the integrated circuit, which are then optically aligned and brought into contact. The solder bumps are approximately 20 x 20 microns, provide adequate mechanical integrity and electrical contact. They have a minimal (but not negligible) effect on the acoustics of this essentially air-back transducer design.

Conventional ultrasound systems use micro-coaxial cable to connect the array to the front end electronics. While coax technology has improved dramatically in the past decade, interconnecting 16,384 array elements with separate wires remains a formidable challenge. In addition to this practical fabrication issue, the capacitance of a long coaxial cable (typically 40 pF/m) is much larger than that of a typical 2D

array element (< 1pF). This creates a voltage divider that severely reduces the signal to noise ratio of the channel. With the flip-chip interconnect method, the total input capacitance is less than 100 fF. This provides a unique opportunity to optimize composite piezoelectric material that is generally not available to conventional ultrasound systems.

Figure 4 plots piezoelectric coupling coefficient, capacitance and sensitivity at *constant resonance* frequency as a function of piezoceramic volume fraction[3]. By reducing the volume fraction of ceramic, sensitivity can be increased at the expense of lower capacitance while maintaining high coupling.

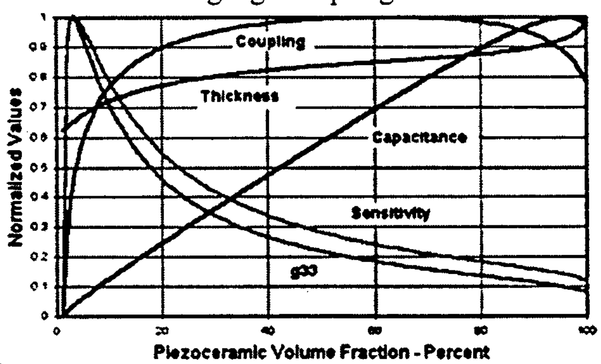

*Fig. 4. Normalized 1-3 Piezoelectric Composite Parameters at **Constant Resonance Frequency** as a function of Volume Fraction.*

Readout Integrated Circuit (ROIC)

Traditional ultrasonic imaging systems, whether for medical or underwater use, are operated in a monostatic mode, i.e., the same array element is used for both transmission and reception. This poses practical difficulties in fabricating an integrated circuit with both the high density required for massively parallel on-chip signal processing and the high voltage capabilities required for adequate transmitted power. The bistatic mode (Figs. 1 & 2), where the transmitter is separated from the receiver was selected for the array discussed here.

From a silicon foundry perspective, matching the silicon unit cell to the full size array dimensions would result in a very large die. To improve yield, a four chip hybrid was selected. To avoid a row of dead pixels in the center of the array, the integrated circuit is designed to be butted along two adjacent edges, with all bond pads on the other two edges. The resulting die is 15.5 mm x 15.5 mm and contains 64 x 64 (4096 total) unit cells together with multiplexing circuits for readout. There are about 1 million active devices per die.

Figure 5 is a schematic diagram of the ROIC. In each unit cell, amplification, filtering, analog sampling and capacitor storage of the output of all the transducer elements in the array occurs simultaneously (in parallel) during data acquisition periods (range gates) which are typically a few microseconds long. After data acquisition, the stored data are read out through four 10 MHz pixel rate outputs.

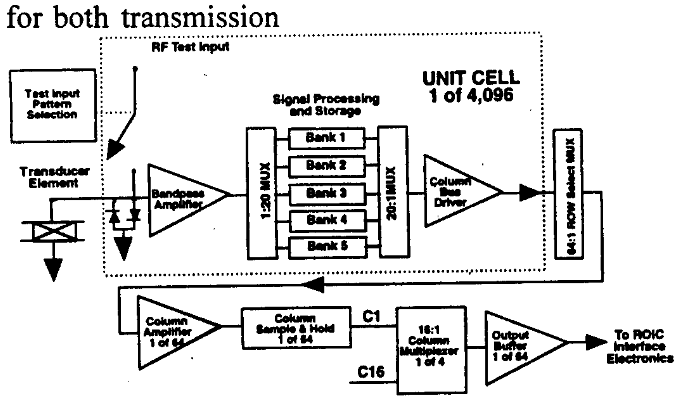

Fig. 5. Functional Block Diagram of the Readout Integrated Circuit (ROIC).

In the sample and hold mode, the received waveform is simply sampled at a 20 Megasample/sec maximum rate when the range gate is activated. In the quadrature detection mode[4], the in-phase and quadrature components of all 16,384 waveforms are simultaneously recorded at 5 separate range gate times. Following data acquisition, the stored data are read serially through four outputs per ROIC for a total of 16 read out channels.

Initial Measured Performance

A 16x16 3.1 MHz sub-array was hybridized to a single ROIC for initial testing. The average measured sensitivity was -188 (±6) dB re 1V per μPa with a broadband electronic noise level of 84 μV. Using relatively uniform direct insonification, typical measured responses are shown in Figs. 6-9. In these figures, the voltage response of each element is plotted as a surface over a 2D grid representing the element location. The lower response in the corners of the array was found to be caused by lower transmission through a matching layer defect. In addition, for this initial test a less than perfect ROIC was used, resulting in some non-functional rows and columns which are omitted from the plots.

With a 60 μsec long gated sine pulse, activity may be seen in each of the 5 planes (Fig. 6). Figure 7 shows a short acoustical pulse, captured in one range plane. Without any acoustical input (Fig. 8), the "fixed pattern" noise of the THA may be observed. Finally, by subtracting two subsequent frames, the fixed pattern noise of the THA is canceled out, leaving only the random noise of the front end electronics (Fig. 9).

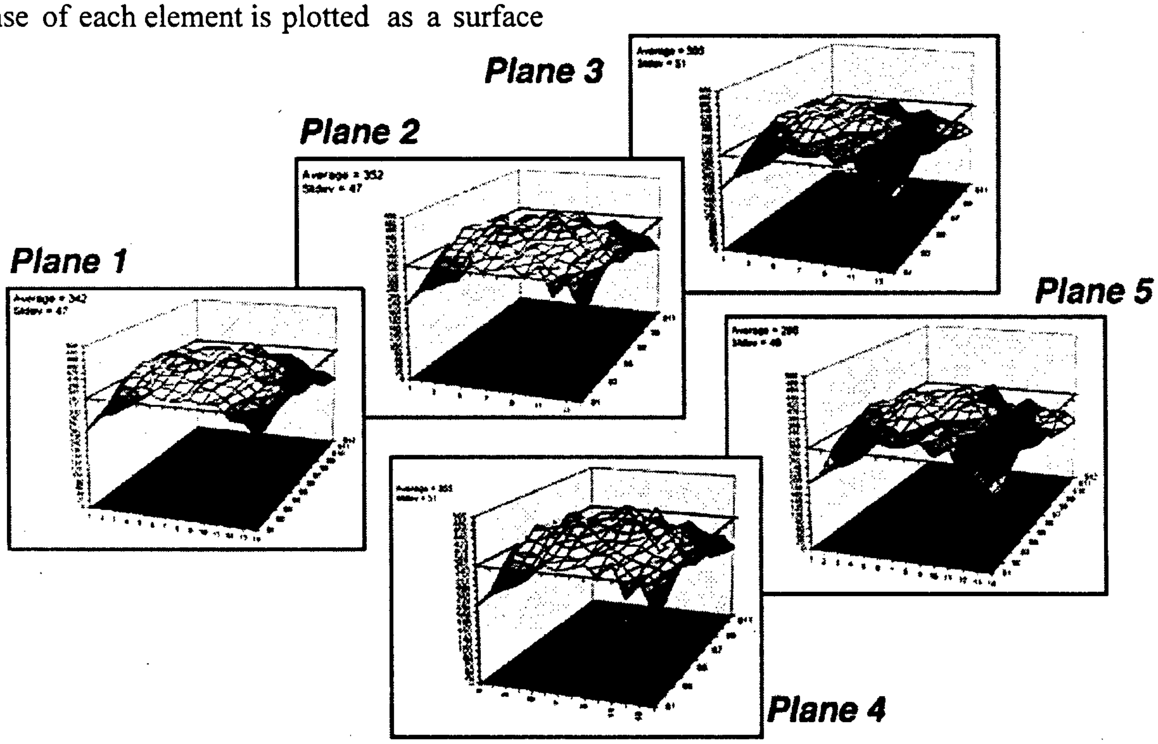

Fig. 6. Response of the test THA to a 60 μsec gated sine pulse of low amplitude ultrasound. The range gates were set to capture the full acoustical pulse.

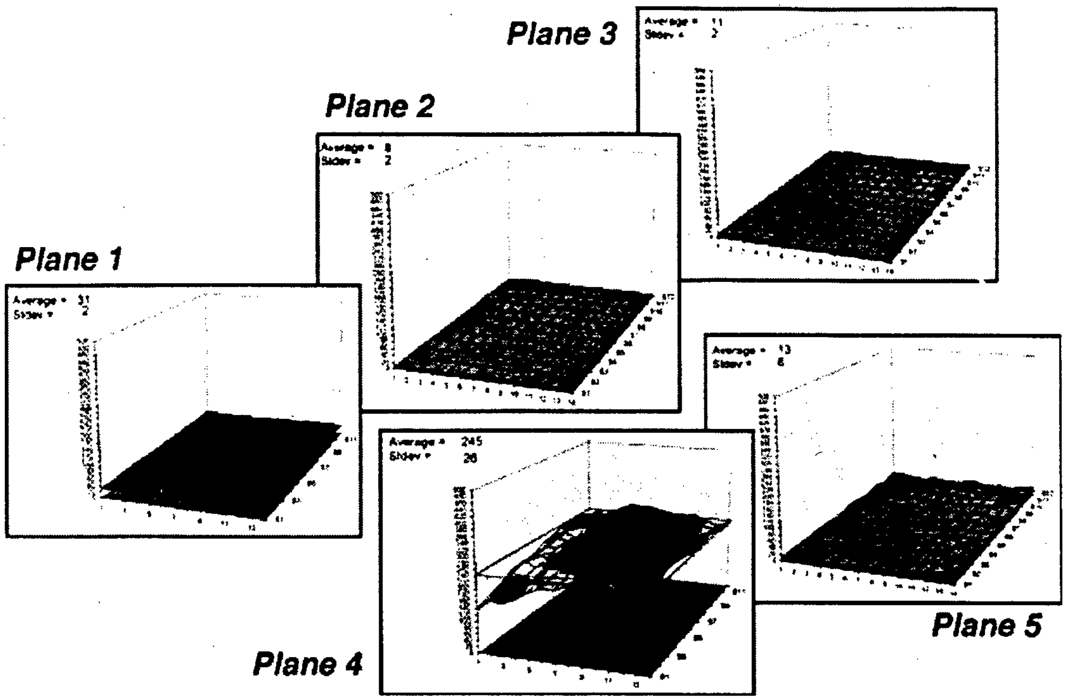

Fig. 7. THA Range resolution. A short acoustical pulse (4.0 μs) appears only in range plane 4. The range gates are 3.25 μs long and are spaced by 0.75 μs, corresponding to pulse-echo ranges of 2.4 mm and 0.5 mm respectively.

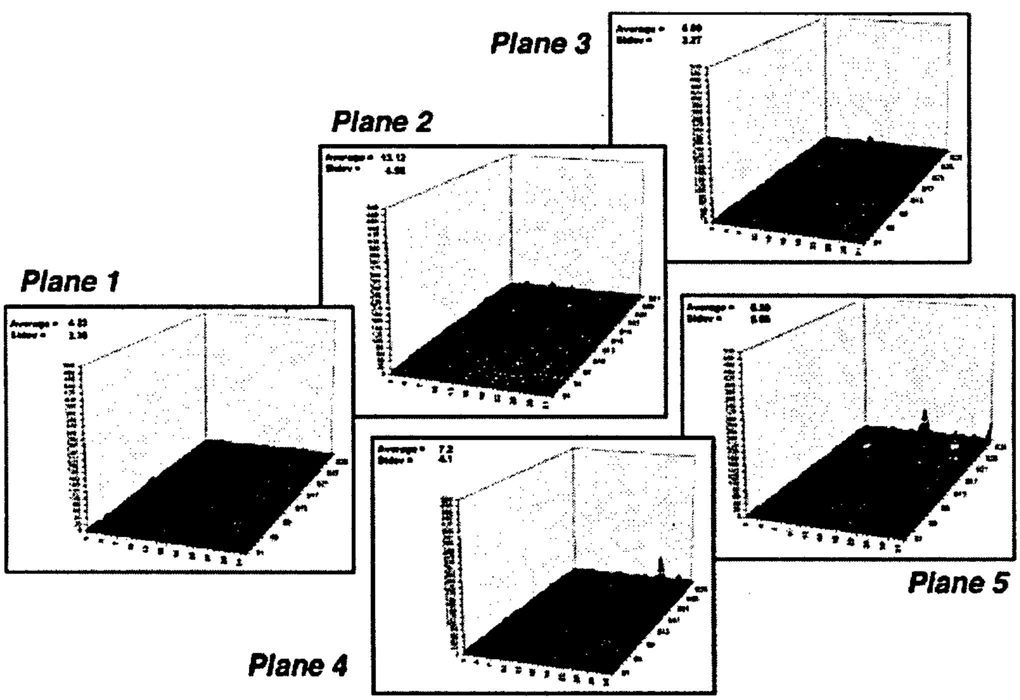

Fig. 8. With no acoustical input, the "fixed pattern" noise of the 16x16 THA is revealed.

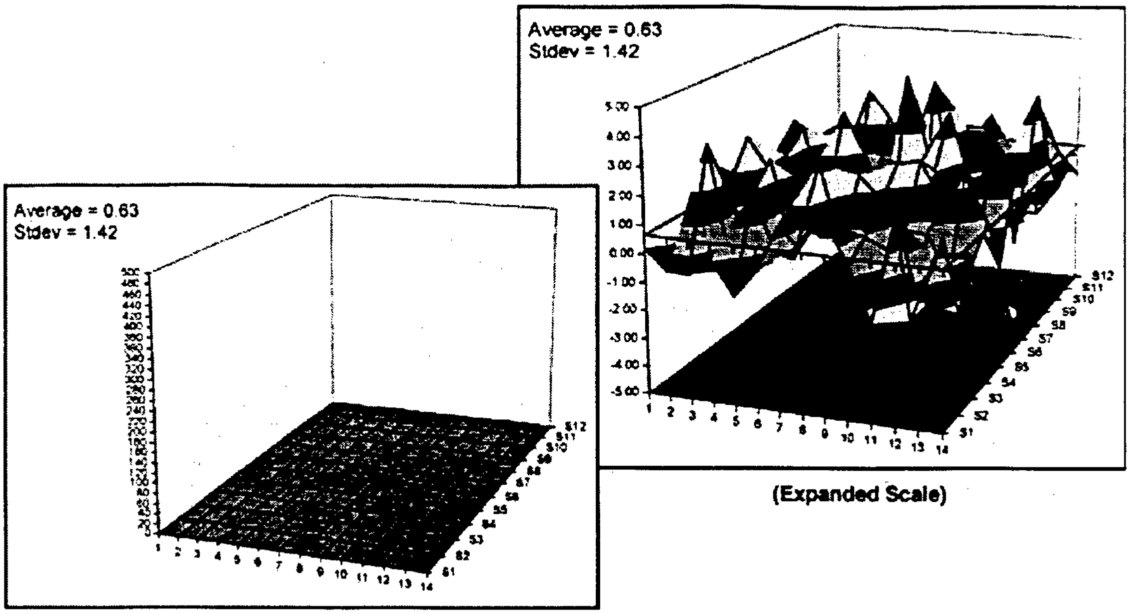

Fig. 9. THA frame to frame repeatability is demonstrated by subtracting the same plane in one frame from the next frame. The difference is equal to the system electronic noise level. The right plot (b) is an expanded version of the left plot (a).

III. CONCLUSION

Fabrication and testing of 64x64 3 MHz and 128x128 5 MHz THA's is underway (Fig. 10).

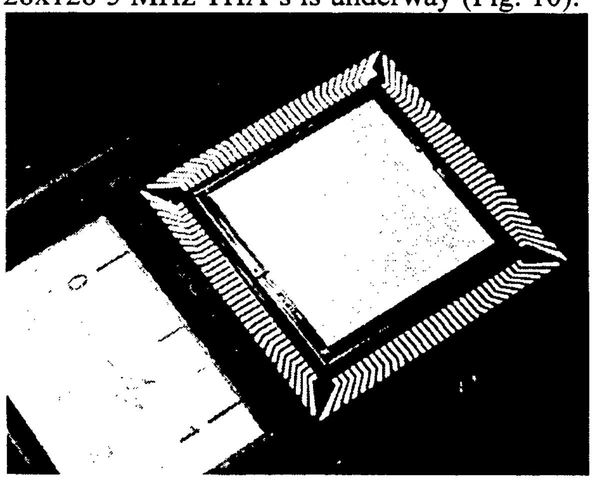

Fig. 10 A 64x64 element 3.1 MHz THA.

We have demonstrated the basic functionality of our THA design and shown that the initial versions meet the system performance goals.

IV. ACKNOWLEDGEMENTS

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V. REFERENCES

[1] K. Erikson and R. Zuleeg, "Integrated Acoustic Array", *Acoustical Holography*, Vol. 7, ed. L. Kessler, Plenum Pub. Corp., pp. 423-444, 1977.

[2] American Medical Design, Paso Robles, CA 93446

[3] W.A. Smith and B.A. Auld: "Modeling 1-3 Composite Piezoelectrics: Thickness-Mode Oscillations", *IEEE Trans. Ultrasonics, Ferroelectrics and Frequency Control*, **UFFC-38:**40-47 (1991).

[4] D. Horvat, J. Bird and M. Goulding, "True time-delay bandpass beamforming", *IEEE J. Oceanic Eng.*, 17:185-192 (1992).